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Use of high wavenumber Raman spectroscopy for measuring tissue

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Use of high wavenumber Raman spectroscopy for measuring tissue

Field of invention

This invention relates to an instrument and the use thereof for measuring a Raman signal of a tissue, comprising a laser, a signal detection unit, and a fiber optic probe.

Background of invention

Atherosclerosis is an important cause of death in many parts of the world. Therefore, many techniques have been developed to obtain information about the plaque that develops in blood vessels. Imaging techniques such as angiography, magnetic resonance imaging, intravascular ultrasound, and optical coherence tomography provide information regarding the location of a plaque or blood vessel obstruction and about the morphology or internal structure of the plaque. However, they do not enable detailed in vivo analysis of the molecular composition of the plaque. Knowledge of the molecular composition of a plaque is important e.g. for determining the risk of acute cardiac events. So-called stable plaque and vulnerable plaque are distinguished, where it is thought that the vulnerable plaque can give rise to such acute, often fatal events. Such an event is triggered by rupturing of the thin fibrous cap of the plaque, bringing the contents of the lipid pool of the plaque into contact with the blood stream, leading to thrombogenesis and occlusion of the artery.

Fluorescence based methods have been shown to be able to distinguish between normal artery wall and atherosclerotic plaque *in vitro*. However fluorescence spectra are easily disturbed by light absorbing molecules in the tissue and in blood, limiting its applicability.

Of all methods to obtain information about atherosclerotic plaque composition and which can in principle be applied *in vivo*, intravascular Raman spectroscopy provides the most detailed information. In Raman spectroscopy, the Stokes-shift between light that is incident on a sample that is investigated and the light that is inelastically scattered by the sample is expressed in relative wavenumbers ($\Delta cm^{-1} = (1/\lambda_{in} - 1/\lambda_{scattered})10^{-2}$ with λ (wavelength) in meter). The wavenumber region between about

400 and 2000 cm^{-1} of the Raman spectrum (the so-called fingerprint region) is used to obtain this information. This region of the spectrum contains many bands that can be discerned and which individually and/or in combinations can be used to obtain information about the molecular composition of the tissue.

Studies in the field of atherosclerosis are only related to the finger-print region, since this spectral region is very informative for analysis or diagnosis. Examples of such studies are e.g. found in the papers of H.P. Buschman, E.T. Marple, M.L. Wach, B. Bennett, T.C. Schut, H.A. Bruining, A.V. Bruschke, A. van der Laarse and G.J. Puppels, Anal.Chem. 72 (2000), 3771-3775, which discusses the *in vivo* determination of the molecular composition of artery wall by intravascular Raman spectroscopy, using a multifiber probe and measuring in the 400-1800 cm^{-1} region; R.H. Clarke, E.B. Hanlon, J.M Isner, H. Brody, Appl. Optics 26 (1987), 3175-3177, which discusses laser Raman spectroscopy of calcified atherosclerotic lesions in cardiovascular tissue, also in the fingerprint region; and J.F. Brennan, T.J. Romer, R.S. Lees, A.M. Tercyak, J.R. Kramer, M.S. Feld, Circulation 96 (1997), 99-105, which deals with the determination of human coronary artery composition by Raman spectroscopy in the fingerprint region.

In vivo application of Raman spectroscopy in most cases requires the use of a flexible light guiding device of small diameter. This can for instance be introduced in the lumen of an artery. It must be able to reach and interrogate locations with atherosclerotic lesions. It can also be used in the working channel of an endoscope or inside a biopsy needle or biopsy forceps. The fiber optic probe (comprising one or more optical fibers) must guide light to the tissue under investigation, collect light that is scattered by the tissue and transport this collected light away from the tissue towards a spectrum analysis device.

Unfortunately, in the 400-2000 cm^{-1} spectral region, the materials of the optical fiber itself generate Raman signal, resulting in a strong signal background. Moreover, bending of the fiber leads to variations in the amount of signal obtained from the core, cladding and coating materials, further complicating signal detection and signal analysis. This deteriorates the signal-to-noise with which the tissue Raman signal can

be detected, and also otherwise complicates signal analysis, and therefore negatively affects clinical utility. It is therefore necessary to use optical filters at or near the distal end of the fiber optic probe which is in contact or in close proximity to the tissue, in order to suppress background signal contributions to the detected Raman signal. However, this in turn necessitates the use of separate optical fibers for guiding laser light to the tissue and for collecting and guiding scattered light away from the tissue. It furthermore often necessitates the use of beam steering arrangements or a lens or lenses at the distal end of the fiber optic probe in order to obtain the desired overlap between the volume of tissue illuminated by the laser light and the volume of tissue from which Raman signal can be collected. Fiber optic probes for Raman spectroscopy are therefore complex. It is difficult to miniaturise fiber optic probes for Raman spectroscopy and to keep them flexible, which is necessary for instance for intravascular use and for use in the auxiliary channel of an endoscope. The complexity is also an obstacle to the production of such probes at a price that they can be used as disposables in hospitals. Moreover, signal intensity of tissue in the 400-2000 cm^{-1} is low, necessitating relatively long signal integration times, which may be impractical for clinical use. All above mentioned problems and disadvantages hinder the actual implementation of Raman spectroscopy for clinical diagnostic purposes in general, and for intravascular use in particular.

US 5,293,872 teaches the use of near-infrared (NIR) laser light excited Raman spectroscopy for distinguishing between normal artery tissue, calcified atherosclerotic plaque and fibrous atherosclerotic plaque. For *in vivo* measurements in the 700-1900 cm^{-1} region, the use of a bundle of optical fibers is discussed. This will lead to the same disadvantages as discussed above, e.g. with respect to noise.

US 5,194,913 recognises the problem of multiple fiber optics, but also notes that the use of a single fiber is prohibited by the fact that background Raman signal generated in the fiber optics is intense for all but the shortest fibers. It discloses a fiber optic apparatus using two opposite fibers and using optical filters to reduce background Raman emission from the fiber optics. This document is related to the problem of signals in fibers in general, and it is clear that the solution provided by US 5,194,913, i.e. an axial configuration, cannot easily be used for measurements *in vivo*.

A paper of J.F. Aust, K.S. Booksh and M.L. Myrick, Applied Spectroscopy 50 (1996), 382-386 discusses cases in which the signal obtained from the sample is relatively strong (polymer) or in which special measures were taken, such as increasing the measurement volume from which sample-Raman signal is obtained, to increase signal intensity from polymers to levels that are very much higher than would be obtained from a biological tissue. This paper does not discuss the applicability of the method to tissue, but teaches that for a good signal, a special teflon tube of up to 4 cm has to be used on the tip of the optical probe, filled with the polymer, in order to get a good signal. Such a method is usually not applicable to tissue, especially not in the case of *in vivo* measurements.

Next to atherosclerosis, cancer is also an import health issue. The same problems as encountered above apply for determining tumor cells by Raman spectroscopy via fiber optics. US 5,261,410 teaches to use a bundle of fibres and to measure in the fingerprint region. Such use also leads to a signal to noise ratio which is not satisfying.

From the above it is clear that there is a need for an instrument for measuring a Raman signal of a tissue, that does not have above mentioned problems.

Summary of invention

The invention is related to the use of an instrument for measuring a Raman signal of tissue, the instrument comprising a laser, a signal detection unit for measuring the Raman signal, and a fiber optic probe, wherein the fiber optic probe comprises one or more optical fibers for directing laser light onto the tissue and for collecting light that is scattered by the tissue and guiding the collected light away from the tissue towards the signal detection unit, wherein the fiber or fibers for collecting light have substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region, and wherein the detection unit records the Raman signal scattered by the tissue in said spectral region.

Brief description of drawings

Figure 1 shows the results of a Raman–mapping experiment (fig. 1 A) in which Raman spectra from a thin section of arterial tissue were obtained in the higher wavenumber

region, enabling identification of tissue areas with different molecular composition, and the correspinding Raman spectra (fig. 1 B).

Figure 2 shows spectra of lipids and proteins that can be present in atherosclerotic plaque and artery wall: A: elastine, B: cholesteryl linoleate, C: cholesteryl oleate, D: cholesteryl linolenate, E: cholesteryl palmitate, F: collagen type 1, G: trilinoleine, H: triolene, I: tripalmitine, J: cholesterol.

Figure 3 provides a comparison of the lipid composition of segments of human arteries as determined by Raman spectroscopy and HPTLC (High Performance Thin Layer Chromatography).

Figure 4 shows the results of a Raman–mapping experiment in which Raman spectra from a thin section of human dura, infiltrated by meningioma (MG), were obtained in the higher wavenumber region, enabling discrimination between these tissues (fig. 4A) and an adjacent H&E (hematoxylin and eosin) stained section (fig. 4B).

Figure 5 shows the results of a Raman–mapping experiment (5A) in which Raman spectra from a thin section of human glioblastoma, were obtained in the higher wavenumber region, enabling identification of areas of vital tumor (V) and of areas of necrosis (N), when compared to figure 5 B, where an adjacent H&E stained section is shown.

Figure 6 schematically shows a set-up for obtaining Raman spectra in the higher wavenumber region.

Figure 7 shows a spectrum (A) of a mixture of lipids measured with a Raman setup according to Figure 6. Also shown are the spectrum of the fiber optic probe itself (B, obtained without a sample present at the distal end of the optical fiber) and a difference spectrum C (A-B).

Figure 8 shows Raman spectra (t) of a normal artery wall (A) and an atherosclerotic artery wall (B), of the results (f) of a least squares fitting of these spectra with the set of

spectra of purified compounds shown in figure 2 and of residuals (r) which represent the signal contained in the tissue spectra that is not accounted for by the set of fit-spectra.

Figure 9 schematically shows an embodiment in which Raman spectroscopy is combined with fluorescence and NIR-absorption spectroscopy.

Detailed description of invention

The invention is related to the use of an instrument for measuring a Raman signal of tissue, the instrument comprising a laser, a signal detection unit for measuring the Raman signal, and a fiber optic probe, wherein the fiber optic probe comprises one or more optical fibers for directing laser light onto the tissue and for collecting light that is scattered by the tissue and guiding the collected light away from the tissue towards the signal detection unit, wherein the fiber or fibers for collecting light have substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region, and wherein the detection unit records the Raman signal scattered by the tissue in said spectral region

The advantage of using this apparatus is that rapid *in vivo* characterisation of tissue, and diseased tissue e.g. atherosclerotic plaque, tumors, pre-cancerous tissue and benign tissue lesions with the Raman spectrometer is enabled and that the signal collection time needed to obtain a Raman spectrum with sufficient signal-to-noise ratio is decreased. A further advantage is that, because no extra means, like e.g. filters are needed that limit the spectral througput of the light guides, Raman measurements can easily be combined with other informative techniques such as fluorescence measurements, near-infrared absorption measurements and optical imaging techniques, which could make use of the same light guides, using the light guides both for guiding light to the tissue as well as collecting light from the tissue and guiding it back to respectively a fluorescence or a near-infrared detection unit or using separate light guides for this.

The invention described herein is based on the surprising finding that very detailed information about the composition and compositional heterogeneity of an

atherosclerotic plaque could be obtained, brain tumor tissue could be distinguished from normal brain tissue and from skull tissue, and necrotic brain (tumor) tissue could be distinguished from vital tumor tissue by recording and analysing Raman spectral maps of thin tissue cross sections using only the 2500-3700 cm^{-1} region of its Raman spectrum.

Before, the Raman finger print region was used to detect these kinds of tissue (see e.g. US 5,293,872; US 5,261,410; Anal.Chem. 72 (2000), 3771-3775; Appl. Optics 26 (1987), 3175-3177; Circulation 96 (1997), 99-105), but it was not known or suggested that above mentioned tissue would also have characteristic and distinctive Raman signals in this higher wavenumber region. Selection of this region has large advantages. For measurements in the finger-print region of the Raman spectrum it is necessary to suppress the intensity of the elastically scattered laser light with special optical filters that combine a deep attenuation of the intensity of the elastically scattered laser light with a high transmission at wavelengths close to the laser wavelength. However, in the present invention a large wavelength shift exists between the incident laser light and the Raman-scattered light in the high wavenumber region. This enables the use of very simple and inexpensive absorption filters in the signal detection pathway for suppression of the intensity of elastically scattered laser light, such as e.g. a colour glass filter.

In general the intensity of the Raman signal of tissue is significantly higher (by a factor of about 5 or more) in this higher wavenumber region than in the 400-2000 cm^{-1} region (fingerprint), enabling reduced signal collection times, e.g. also about by a factor of about 5 or more.

Another advantage of selecting this region is that this enables the recording of tissue Raman signal, using a single optical fiber to illuminate the tissue and to collect Raman signal from the tissue, with the tissue Raman signal being of comparable intensity or even higher intensity than that of the background signal generated in the optical fiber. Some fibers are very suitable for these kind of measurements, since the Raman scattering of the fiber itself in this wavelength region is low or negligible compared to the signal of the tissue. This is different from the fingerprint region where in the same

configuration the signal background of the optical fiber, in practical situations, using a fiber of several meters in length, has an intensity that is usually more than an order of magnitude higher than the Raman signal of the tissue. In addition, the background signal from some types of optical fiber in the 2500-3700 cm^{-1} region consists of only a signal of which the intensity variations as a function of wavenumber shift are very small compared to that of the tissue Raman signal and therefore can be easily distinguished from the tissue Raman signal and/or accounted for in the signal analysis. In the fingerprint region the background signal from the fiber has sharper features making signal analysis harder. Hence, the signal-to-background ratio in the wavelength region of the invention is much higher, than in the fingerprint region. This is due to the finding that the Raman signal of the fiber is absent or strongly diminished in the higher wavenumber region of the Raman spectrum, whereas in the fingerprint region, as used in the prior art, the fiber also generates a Raman signal, disturbing or even overcoming the Raman signal of tissue or sample.

Especially the 2700-3100 cm^{-1} region of the Raman spectrum is informative for the above mentioned tissue. Therefore, in a preferred embodiment of the invention, the detection unit of the instrument records the Raman signal in one or more parts of the 2700-3100 cm^{-1} spectral region. A further advantage therefore is that now, only signal in a small wavenumber region needs to be recorded, enabling the use of a multichannel light detector with fewer channels. Though the 2700-3100 cm^{-1} spectral region is especially informative for detecting, analysis and diagnosis of deseases in tissue, preferablly atherosclerotic plaque and cancerous or pre-cancerous tissue, the invention does not exclude measurements outside the mentioned spectral regions in order to obtain additional information. The invention also comprises in an embodiment collecting Raman signal in other spectral regions (e.g. the fingerprint region) in addition to the 2700-3100 cm^{-1} spectral region.

An advantage of the use of the Raman generation and detection instrument is that the complexity of Raman spectrometers for measuring samples (especially (in vivo) tissue measurements), characterisation and/or tissue classification, is decreased by using the higher wavenumber spectral region and carefully selecting light guides that serve both to guide laser light to the tissue as well as to guide away light that is scattered by the

tissue. Hence, the invention comprises the use of an instrument, wherein the fiber optic probe comprises an optical fiber that not only directs laser light onto the tissue but also both collects light that is scattered by the tissue and guides the collected light away from the tissue towards the signal detection unit. This embodiment also encompasses a fiber optic probe having a number of fibers that serve both to guide laser light to the tissue as well as to guide away light that is scattered by the tissue. Since one, a number, or all of such fibers in the probe can do this, the dimensions of the fiber optic probe can be diminished with respect to state of the art fiber optic probes for tissue characterization (which comprise different fibers for guiding laser light to the sample and detecting the Raman signal).

A further advantage is that the size of a Raman catheter for in vivo intravascular use can even be minimised to only one single optical fiber. This means that the diameter of e.g. the intravascular fiber optic probe can be maximally reduced and that maximal fiber optic probe flexibility can be achieved, which are also highly desirable attributes of catheters for intravascular use. Also for other applications, where small fiber are desirable, the instrument can be used.

A significant reduction of complexity and following from that a reduction of production cost of the fiber optic is another advantage. The fiber could even be used as a disposable, which is highly desirable for an intravascular catheter in clinical use.

In another embodiment of the invention, the Raman measurements can be combined with fluorescence and/or near-infrared absorption measurements. Hence, the detection unit will also comprise a detector for measuring fluorescence and/or a detector for near-infrared absorption. In this embodiment, it is e.g. possible that the fluorescence and/or near-infrared absorption measurements make use of a fiber also used in obtaining Raman signal.

In a further embodiment, only one single optical fiber is used for directing laser light (and (N)IR light) onto the tissue, as well as for collecting Raman signal that is scattered by the tissue, for collecting fluorescence and/or near-infrared signal, and for guiding

the collected light away from the tissue towards the signal detection unit, which comprises the respective detectors.

In another embodiment, a plurality of fibers can be used to get an enhanced signal. This embodiment also comprises the use of an instrument wherein Raman measurements can be combined with fluorescence and/or near-infrared absorption measurements and wherein the detection unit also comprises a detector for measuring fluorescence and/or a detector for near-infrared absorption. In a another embodiment, the dimensions of the probe can further be diminished, when fluorescence and/or near-infrared absorption measurements make use of a fiber also used in obtaining Raman signal. Here, 'a fiber' encompasses again one or more fibers. Such bundles of fibers can be used for measuring and/or scanning a tissue area. The advantage is that measurement locations can be closer together than in state of the art fiber optic probes, raising the resolution.

The small diameter and high flexibility provide the best possibilities for combining the Raman probe with other sensing modalities (e.g. intravascular ultrasound for intravascular use, and incorporation in an endoscope for oncological applications) and for incorporation in instruments for obtaining tissue samples (such as a biopsy forceps or a an instrument for obtaining fine needle aspirates) or with treatment modalities (e.g. devices that use heat to coagulate tissue, such as tumor tissue, or surgical instruments). Hence, the invention also comprises an instrument wherein part of the fiber is integrated or combined with a catheter that provides additional information about the tissue or which comprises means to obtain tissue samples, means to treat tissue and/or means used in surgical procedures.

All these advantages allow a much simplified instrumentation. The instrument therefore enables *in vitro* and *in vivo* analysis and diagnosis of atherosclerotic plaque and detection of tumor tissue with great advantages over current state-of-the-art technology.

In the context of this invention, "tissue" refers to tissue of human, animal or plant origin, but in particular is meant human or animal tissue. Tissue includes a biological cell or cells, an organ, or part, of a human or animal body or a substance extracted from, or from a part of, the human or animal body and can e.g. be bone, blood or brain.

Tissue samples can be measured, i.e. Raman signals can be measured, which are elicited by illumination with light emitted by the laser, *in vitro* or *in vivo*. Tissue is considered to belong to a particular clinical diagnostic class if it possesses one or more characteristic features, which may include but are not limited to, morphological, chemical, and genetic features. These can be typical of a certain pathological condition.

The fiber tip can be in or on the tissue, but can also be in close proximitiy, e.g. a few mm. However, the proximity can also be larger, when a lense is used to image the distal end of the fiber onto the tissue. In some cases, the tip cannot be on the sample, e.g. when the sample is measured through for example glas. In such a case, the proximity can even be a few centimeters or more. Proximity in this invention comprises both above-mentioned options.

The laser in this invention is any monochromatic light source with a line width sufficiently narrow to enable measurement of the desired Raman signal of a sample with sufficient spectral resolution, like a laser. The line width will in most cases preferably be below 5 cm^{-1}. The light beam of such a source is coupled into a fiber, and the light is shed on a sample. A Raman signal of such a sample may be produced by illuminating it with light from such laser source, provided that the sample contains molecules that have molecular vibrational modes that can participate in Raman scattering of incident light. Preferably for Raman measurements of tissue, the laser or source has an emission above about 600 nm, since in this way absorption of incident laser light in tissue is minimised and also autofluorescence of tissue is minimised. Autofluorescence can cause a background signal to the Raman spectrum which deteriorates the signal-to-noise with which the Raman signal is detected. Examples of sources are e.g. diode lasers, He-Ne lasers Ti-sapphire laser etc.

With "instrument" in the invention is meant a spectrometer comprising a combination of a laser, for producing a Raman signal, an optical fiber and a signal detection unit.

The spectrometer may comprise a filter to suppress the intensity of the component of the light that is guided to the spectrometer that has the same wavelength as the laser light. This filter should suppress the intensity of this light by preferably 8 orders of

magnitude or more, while suppressing the intensity of the Raman scattered light in the wavenumber region of interest by preferably less than 10%. Because the higher wavenumber spectral region is used, implying a large wavelength interval between the laser light and the wavenumber region of interest, this may be a simple colour glass absorption filter, such as e.g. RG 780 colour glass filter from Schott. Two of such filters in series and of 3 mm thickness (both commercially available) will suppress laser light below 725 nm by 10 orders of magnitude or more, while causing no significant attenuation of the Raman signal of interest other than reflection losses at glass air interfaces. Preferably, the entrance and exit faces of the filter are coated with an anti-reflection coating optimised for the wavelength region of interest, so as to minimise reflection losses at air-glass interfaces. In that way a throughput of Raman signal of more than 90% can easily be achieved. The spectrometer preferably has no moving parts, is optimised for throughput in the NIR, and has a resolution of preferably at least 8 cm^{-1}.

However, in some cases fluorescence might be desired (see above), as a source of information for characterisation of the tissue and to measure simultaneously or sequentially, either with one or with several fibers the Raman signal and fluorescence of the sample. In such an embodiment, the fluorescence excitation light may have wavelengths below 600 nm, e.g. in the blue or UV.

The signal detection unit preferably comprises detectors like a multichannel CCD-detector optimised for light detection in the NIR. An example of such a detector is a deep-depletion back-illuminated CCD-camera (DU401-BRDD) from Andor-technology (Belfast, Northern-Ireland). The spectral region of interest can e.g. be chosen by a grating or prism. Recorded spectra are preferably displayed and/or analysed by means of dedicated software and a personal computer in real time.

In the context of this invention an optical fiber is defined as a device with a proximal end and a distal end, which is able to guide light from the proximal end to the distal end. The term "a fiber" comprises one or more fibers. The term "fiber optic probe" comprises one optical fiber or a bundle of optical fibers.

The distal of the fiber probe end may be shaped or be fitted with a micro-optical component physically attached to it, to arrive at certain illumination directions and/or angles and/or to arrive at certain light collection directions and/or angles and/or to determine the sample surface which is illuminated and/or to determine the size and/or location of the sample volume from which Raman signal is preferentially detected. In the art of measuring tissue with Raman spectroscopy, these probes usually contain one fiber for excitation and at least one fiber, but usually a number of fibers, to guide the (Raman) signal to a detector.

It was also found that some fibers are very suitable for these kinds of measurements because the Raman scattering of the fiber itself in this wavelength region is low or negligible compared to the signal of the sample. Hence, the instrument comprises a fiber which has substantially no Raman signal in the spectral region where Raman signals are found. An example of a preferred fibre is a fiber having a fused silica core and a fused silica cladding, like e.g. WF200/220A optical fiber from Ceramoptec Industries Inc or FG-200-LCR fiber from 3M Company or equivalent fibers. Some fibers are less preferred, like e.g. WF200/220N optical fiber from Ceramoptec Industries Inc or FT-200-EMR optical fiber from 3M Company, which appear to have a large background signal in the spectral region of interest.

Good results are obtained when so-called low OH-fibers were used with a fused silica core and a fused silica or TECS cladding (which have high transmission in the near-infrared), and low background signal contributions are obtained with coating materials that have a lower relative refractive index than the core and the cladding and/or show little or no signal in the 2500-3700 cm^{-1} wavenumber interval, such as, but not limited to TECS (material used by 3M company), Tefzel (ethylene tetra-fluoroethylene) and acrylate. Non-limiting examples of suitable fibers are FG-200-LCR (which is a fiber with a fused silica core (200 micron in diameter), a fused silica cladding of 240 micron in diameter, a TECS coating of 260 micron in diemater and a Tefzel buffer of 400 microns in diameter), FT-200-EMT (also from 3M Company) which is a fiber with a cladding made of TECS, and WF 200/240 A, which is a fused silica core/fused silica cladding fiber with an acrylate coating.(from Ceramoptec).

Fibers having coating materials which exhibit a Raman signal in 2500-3700 cm^{-1} interval and which have a higher refractive index than the cladding and the core of the fiber are less preferred, as it was found that Raman light generated in such coatings can get trapped in the fiber core and can then be guided in the fiber core toward the signal detector; examples are silicone coatings (which give rise to a strong Raman signal in the 2700-3100 cm^{-1} spectral interval. and polyimide coatings (which give rise to a high fluorescence background when 720nm laser light is used). Examples of fibers that give rise to unfavourable intense background signal in the 2500-3700 cm-1 spectral interval are WF 200/240 BN and WF200/240 BT, which are fibers with a fused silica core and a fused silica cladding and a silicone buffer with a black nylon respectively a black Tefzel coating (Ceramoptec). Another example is WF200/220 P (from Ceramoptec), which is a fiber with a fused silica core, a fused silica cladding and a polyimide coating.

It is evident that the fiber has also sufficient transmission for the laser light and for the Raman signal of interest. A preferred fiber has a transmission for the wavelengths of the laser light and the Raman signal of at least 50%, more preferably 70%.

In another embodiment, the instrument is an instrument comprising an optical element at the distal end of the fiber optic probe for purposes of defining the location and/or volume of the sample which is illuminated and/or from which scattered light is collected.

With plaque or atherosclerotic plaque in this invention is meant a pathologic condition comprising a build up of fatty materials in the lining of an artery. It may be present in any artery of the body, most frequently in the coronary territory the carotids, aorta, renal arteries, and distal arteries in the legs. Plaque or atherosclerotic plaque in and/or on tissue shows one or more characteristic Raman signals in the 2500-3700 cm^{-1} spectral region. Such Raman signals are especially found around in the spectral region between 2700 and 3100 cm^{-1}.

In a preferred embodiment, the instrument comprises a fiber which has substantially no Raman signal in the spectral region where Raman signals are found which are

characteristic to atherosclerotic plaque. Such Raman signals are especially found in the spectral region between 2700 and 3100 cm^{-1}. This also comprises an instrument, wherein the fiber has substantially no Raman signal in the spectral region where Raman signals are found which are characteristic to one or more of the group of lipid pools, fibrous cap and/or the presence of macrophages or cholesterol therein. The positions of the Raman signals of these compounds can be derived by a person skilled in the art by comparing Raman spectra of tissue that is healthy and tissue that is affected and/or contains such compounds. With substantially "no Raman signal in one or more parts of a spectral region" is meant that the intensity of the detected background signal generated in the fiber is of the same order of magnitude as the Raman signal of the sample under investigation, or lower, in at least part of the spectral interval in which characterising Raman signal is found, and that the Raman signal(s) of the sample can be easily distinguished from this background signal. The instrument can measure in the complete spectral region between 2500-3700 cm^{-1}, preferably 2700 and 3100 cm^{-1}, but it is also possible to select part or parts of this spectral region for measuremens and analysis and/or diagnosis.

With "cancerous tissue" is meant tissue that comprises cancer cells. Pre-cancerous tissue is to be understood as tissue that is abnormal tissue which is a pre-cursor of cancerous tissue. In one embodiment, the instrument has a fiber which has substantially no Raman signal in the spectral region where Raman signals are found which are characteristic to cancerous tissue or pre-cancerous tissue, especially brain cancer. Such Raman signals are found in the 2500-3700 cm^{-1} spectral region, especially in the spectral region between 2700 and 3100 cm^{-1}.

Usually, in order to enable quick and/or automatic analysis, the instrument further comprises a signal analysis unit which analyses the recorded Raman signal. The analysis comprises an algorithm which outputs data regarding e.g. the molecular composition of the sample and/or the clinical diagnostic class to which the sample belongs.

Determination of the molecular composition of e.g. vascular wall or atherosclerotic plaque is accomplished by e.g. a least squares fit procedure in which the measured

spectrum is fitted with a set of spectra of compounds known to be potentially present in the vascular wall or plaque. Quantitative information regarding molecular composition is then obtained from the fit coefficients. Alternatively, e.g. a partial least squares-algorithm may be developed that will accurately determine molecular composition. For detection of cancerous tissue various well known multivariate statistical analysis and/or neural network analysis methods can be employed, such as linear discriminant analysis and artificial neural networks. These analysis and/or diagnostic methods are known in the art, but the specific parameters will be adapted to the respective tissue or sample under investigation.

Such an instrument, comprising a signal analysis unit, is very suitable for use in the diagnosis of diseases, like atherosclerotic plaque and/or cancerous tissue or pre-cancerous tissue. The signal analysis unit can provide information about the molecular composition of normal and atherosclerotic blood vessel wall, the clinical diagnostic class of an atherosclerotic lesion, fibrous cap thickness, the presence of macrophages in the fibrous cap, the presence, size and/or composition of a lipid pool, the presence of cholesterol (esters), the presence of cancerous or pre-cancerous tissue, vital tumor or necrosis, and can provide specific signals for one or more of each.

The invention is also directed to the use of the instrument for measuring a Raman signal of a tissue sample which is isolated before measureing, e.g. excised, biopted tissue, preferably excised, biopted or taken from a human or animal body.

In another aspect of the invention, it comprises an instrument for measuring a Raman signal of tissue, the instrument comprising a laser, a signal detection unit for measuring the Raman signal, and a fiber optic probe, wherein the fiber optic probe comprises one or more optical fibers for directing laser light onto the tissue and for collecting light that is scattered by the tissue and guiding this collected light away from the tissue towards the signal detection unit, wherein the fiber or fibers for collecting light have substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region, and wherein the detection unit is able to record the Raman signal scattered by the tissue.

In one embodiment, the fiber optic probe comprises an optical fiber that both directs laser light onto the tissue and collects light that is scattered by the tissue and guides this collected light away from the tissue towards the signal detection unit, and wherein the fiber has substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region.

In a further aspect of the invention, it comprises a method for measuring a Raman signal of a tissue sample, wherein an instrument according to the invention is used and wherein the tissue sample is excised, biopted or taken from a human or animal body before measuring.

The invention also comprises a method for producing and measuring a Raman signal, comprising sending laser light through an optical fiber, receiving the Raman signal through an optical fiber and detecting the Raman signal by a signal detection unit, characterised by sending the laser light through a same optical fiber which also receives the Raman signal, using an optical fiber for this method which has substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region, and wherein the signal detection unit measures the Raman signal in said spectral region. The end of said optical fiber, which is used to shed laser light on a sample, can be brought in, or in contact with, or in close proximity to said sample. Samples which are Raman active, will give a Raman signal, that can be detected via the same fiber which was used to produce the Raman signal.

In a specific embodiment, the above mentioned method is a method for analysing tissue by measuring a Raman signal, comprising sending laser light through one end of an optical fiber, bringing the other end of said optical fiber in, or in contact with, or in close proximity to the tissue of interest, receiving the Raman signal scattered by the sample through an optical fiber and detecting the Raman signal by a signal detection unit, characterised by sending the laser light through the same optical fiber which also receives the Raman signal, and using an optical fiber for this method which has substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region. If necessary, e.g. to improve the signal to noise ratio, multiple Raman measurements of the tissue under investigation are made.

In another embodiment of the method of the invention, the signal of a detection unit is send to a signal analysis unit which analyses recorded Raman signal, the analysis unit comprising an algorithm which outputs data regarding the molecular composition of the sample and/or the clinical diagnostic class to which the sample belongs.

In order to analyse or make a diagnosis, several methods can be used to derive information. For example, the invention comprises a method, wherein prior to obtaining measurements of the tissue area of interest, measurements are made of tissues normally encountered in the area of interest. But it also comprises a method wherein prior to scanning the tissue area of interest, measurements are made of tissue or tissues affected by the specific disease to be detected in the tissue area of interest and in the same spectral region or a part or parts of this region. Hence, it comprises a method for evaluation of the Raman signal obtained from the tissue region of interest, in order to determine whether such Raman signal was obtained from normal tissue or from diseased tissue.

The method of the invention can be used for diagnosing human or animal blood vessel wall tissue, for diagnosing human or animal tissue on the presence of dysplasia, for determining the molecular composition of normal and atherosclerotic blood vessel wall, for determining the clinical diagnostic class of an atherosclerotic lesion, fibrous cap thickness, the presence of macrophages in the fibrous cap, the presence, size and/or composition of a lipid pool, the presence of cholesterol (esters), the presence of anomalous, cancerous or pre-cancerous tissue vital tumor or necrosis.

The method of the invention can also be used for evaluating the effect of medicaments, food or dietary food, or therapy on diseased or healthy tissue.

Examples

Example 1: Raman mapping of atherosclerotic artery

This experiment describes the possibilities of Raman spectroscopy in the spectral region of the invention for studying artherosclerotic plaque.

The human coronary artery sample used to create the Raman map shown in figure 1 was obtained at the time of autopsy (less than 24 hour *post mortem*). It was snap frozen in liquid nitrogen and stored at –80°C until use. For the Raman measurements a 20 µm thick cryosection was placed on a calcium fluoride (CaF_2) window (Crystran UK) and passively warmed to reach room temperature. After Raman measurements it was stained with a standard hematoxylin and eosin staining procedure.

To collect Raman spectra, 719 nm laser light from an Argon-ion pumped Titanium: Sapphire laser system (Spectra Physics, Mountain View, CA) was used. The Raman microspectrometer system that was used has been described in detail in Van de Poll SWE, Bakker Schut TC, Van der Laarse A, Puppels GJ *"In Situ Investigation of the Chemical Composition of Ceroid in Human atherosclerosis by Raman Spectroscopy"*. J. Raman spectrosc. 33:544-551 (2002). A 80x NIR optimized objective (Olympus MIR-plan 80x/0.75, Japan) with a working distance of approximately 1.6 mm was used to focus the laser light onto the arterial section, and to collect light that was scattered by the tissue sample. For automatic scanning of the tissue sections, the microscope was equipped with a motorized, computer controlled sample stage. The pixel area was scanned through the laser focus in both lateral directions during each measurement, in order to obtain an average Raman spectrum of the entire pixel. Acquisition of Raman spectra and microscopic stage movement was controlled by Grams/32 Spectral Notebase Software (Galactic Industries Corp., Salem, NH). Laser power underneath the microscope objective was approximately 40 mW.

The CaF_2 window with the tissue was placed underneath the microscope. The computer controlled sample stage was moved over a two-dimensional grid, and Raman spectra were acquired with a collection time of 1 second per grid point.

Wavenumber calibration of the spectra was performed using three known Raman calibration standards (4-acetamidophenol, (Sigma), naphthalene, cyclohexane (ICN Biochemicals)), and the emission lines of a neon and a neon-argon lamp. The spectra were corrected for cosmic rays and corrected for wavenumber dependent signal detection efficiency of the setup using a calibrated tungsten band lamp of a known

temperature. Subsequently, the background signal, originating from the optical elements in the laser light delivery pathway was subtracted.

Raman data processing

For all data processing, Matlab 6.1 version R12 (Mathworks Inc., Natick, MA) was used.

K-means cluster analysis

Principal component analysis (PCA) followed by K-means clustering analysis (KCA) was used to determine the heterogeneity in Raman spectra within each tissue sample, in a non-subjective way and without assuming prior knowledge of the morphology and composition of the artery samples. This clustering analysis algorithm was used to find groups of spectra with similar spectral characteristics (clusters). In brief, the analysis was carried out on normalized first derivatives of the spectra (2700 to 3100 cm^{-1}) in order to diminish any influence of variations in the absolute intensity of the Raman signal and to correct for a slight slowly varying signal background due to slight autofluorescence from the tissue. First, PCA was performed on the Raman spectra, to orthogonalize and reduce the number of parameters needed to represent the variance in the spectral data set. The first 100 principal components were calculated, typically accounting for up to 99% of the signal variance. The PC scores, obtained for each spectrum, were used as input for KCA. The number of clusters in which the spectra are grouped by KCA is defined by the user. After KCA, a particular grey-tone was assigned to each cluster. Each grid element of the Raman map was then assigned the grey-tone of the particular cluster to which its spectrum belonged. In this way a grey-tones-image of the frozen section was created, in which areas with similar spectra had the same grey-tone. Finally, the averaged Raman spectrum of each cluster was calculated.

Figure 1 shows the result of a Raman mapping experiment in which spectra were obtained of a thin tissue cross section of unfixed human atherosclerotic artery in a 2-dimensional grid of 80 x 70 points. The differences between spectra obtained from grid points with equal gray tone were smaller than between spectra obtained from grid points with different gray tone, as determined by a K-means cluster analysis of the data.

Tissue grid points with equal gray tone therefore have similar molecular composition. Tissue grid points with different gray tone show significant difference in molecular composition.

A) Result of a 4 clusters K-means clustering analysis. Cluster 1 coincides with adventitial fat. Cluster 2 coincides with artery wall. Clusters 3 and 4 coincide with an atherosclerotic lesion.
B) Cluster-averaged Raman spectra for clusters 1, 2, 3 and 4.

The differences in the spectra of figure 1B, as well as the highly structured localisation of tissue grid points with very similar spectra (belonging to a cluster) illustrate the sensitivity of high wavenumber Raman spectroscopy to the architecture of an atherosclerotic plaque in terms of its molecular composition. From the spectra information about the molecular composition of tissue grid points can be deduced by e.g. a classical least squares fitting procedure, in which tissue spectra are fitted with spectra of e.g. isolated compounds that can be present in the tissue.

Figure 2 shows spectra of such compounds: A: elastine, B: cholesteryl linoleate, C: cholesteryl oleate, D: cholesteryl linolenate, E: cholesteryl palmitate, F: collagen type 1, G: trilinoleine, H: triolene, I: tripalmitine, J: cholesterol. This figure shows that these compounds, which can be present in artherosclerotic plaque and artery wall, have distinctive Raman signals in the spectral region of interest. Raman spectra of these chemicals were recorded using the same Raman setup as used for the measurements shown in figure 1.

Table 1 shows the result of a least squares fit of cluster averaged spectra 1-4 of figures 1B with the pure compound spectra of figure 2 and a 1st order polynomial to account for a slightly sloping background. Cluster averaged Raman spectra were fitted with the set of Raman spectra of these pure compounds, using a non-negative (which means that only positive fit-coefficients are allowed) linear least squares fitting routine. The first order polynomial was included in the fit to account for a slight (fluorescent) background in the Raman spectra. The sum of the non-negative least squares fit contributions of the compound spectra was set to 100%.

The percentages shown relate to the relative signal contributions of the protein, cholesterol, triglyceride and cholesterol ester spectra shown in figure 2. Signal contributions of different cholesterol esters were co-added ("total cholesterol esters" in table 1), signal contributions of different triglycerides were co-added ("total triglycerides") as well as those of collagen and elastin ("total proteins").

TABLE 1: Relative signal contributions of cholesterol, cholesterol-esters, triglycerides and proteins signal obtained from different regions of an artery wall containing an atherosclerotic lesion.

Cluster	Location	Cholesterol	Total cholesterol esters	Total triglycerides	Total protein
1	adventitial fat	0%	11%	88%	1%
2	Normal artery wall	2%	0%	0%	98%
3	Athero-sclerotic lesion	14%	33%	29%	24%
4	artery wall surrounding lesion	2%	21%	3%	73%

Figure 3 shows the result of a comparison of the lipid composition of human arterial segments as determined by Raman spectroscopy and by HPTLC (high-performance thin-layer chromatography)). 58 arterial segments of ~1 cm^2 were scanned under a Raman microspectrometer while Raman signal was collected in the higher wavenumber region (same instrument as for figures 1 & 2). After the Raman measurements, lipids were extracted from the arterial segments and analysed by means of HPTLC. Total lipid fraction was normalised to 100%. A partial least squares analysis model was developed based on Raman and HPTLC results of 57 segments and applied to the Raman spectrum 58th segment to predict its lipid composition. The outcome was

compared with the HPTLC analysis of the 58th segment. This leave one out evaluation was repeated for each of the 58 segments. Figure 3 shows a comparison of the higher wavenumber Raman method for lipid composition determination in human arteries (in situ) and HPTLC for relative weight fractions cholesterol, total cholesterol esters and total triglycerides. High correlation coefficients were obtained (r=0.95 for cholesterol, r=0.93 for cholesteryl esters, r=0.96 for tryglycerides).

This experiment shows that Raman measurements in the spectral region of the invention give very good results and comparable information to HPTLC, enabling Raman spectroscopy as in vivo technique for studying artherosclerotic plaque.

Example 2: Raman mapping of cancerous tissue

This experiment describes the possibilities of Raman spectroscopy in the spectral region of the invention for studying cancerous tissue.

The high wavenumber region can also be used advantageously in various clinical oncology applications. For instance, figure 4A shows a Raman map obtained of a thin tissue section of human dura infiltrated by meningioma in a way similar to the map of an atherosclerotic lesion in figure 1A. Currently no good intra-operative assessment of excision margins is possible. However, it is known that meningioma tissue that is left behind may lead to recurrence of the tumor. Figure 4B shows a picture of an adjacent tissue section after staining with hematoxylin and eosin (H&E stained). Surprisingly, the histopathological evaluation of this section and its comparison with the Raman map show that the light gray areas in the Raman map correspond to dura, while the dark gray areas correspond to meningioma (MG).

This experiment shows that Raman measurements in the spectral region of the invention give valuable information on cancerous tissue of the brain, enabling Raman spectroscopy as in vivo technique for studying such tissue.

Example 3: Raman mapping of cancerous tissue

This experiments describes the possibilities of Raman spectroscopy in the spectral region of the invention for studying cancerous tissue.

Figure 5A shows a Raman map of a thin section of human glioblastoma with both vital tumor areas and areas with necrotic tissue. Surprisingly, comparison of the Raman map with the H&E stained adjacent section evaluated by a neuropathologist, shows that the light gray area corresponds to vital tumor tissue while the dark gray area in the Raman map corresponds to necrosis.

This experiment shows that Raman measurements in the spectral region of the invention give valuable information on cancerous tissue of the brain, enabling Raman spectroscopy for discriminating between vital tumor tissue and necrosis.

Example 4: Schematic representation of Raman spectrometer

Figure 6 schematically shows a characteristic Raman measurement and analysis set-up comprising, a laser 100, coupling optics 110, by which laser light following a first light path 105 is coupled into a fiber optic probe 120, which guides the laser light to the tissue 130 under investigation and which collects light scattered by the tissue and guides it back to coupling optics 110, a filter 140 which creates a light path 145 for Raman scattered light from the tissue 130, which is shifted in wavelength with respect to laser light from laser 100, a filter 150 for strong attenuation of remaining light of the same wavelength as the laser light in light path 145, a measuring unit 160, which measures intensities of the Raman scattered light at a plurality of wavelengths, a signal storage device 170 which may be electronically linked to measuring unit 160 and which stores measured intensities, and a signal analysis device 180, which may or may not be physically linked to signal storage device 170 or which may coincide with signal storage device 170, and which analyses the measured signals for instance to provide information about the molecular composition of the tissue 130 or to enable classification of the tissue, e.g. determination of the clinical diagnostic class to which the tissue belongs. The system can comprise a unit that gives an audible or visible signal when certain tissue is encountered. The invention is not limited to this configuration; the person skilled in the art can vary and/or choose the components which are according to his knowledge desirable or necessary.

Example 5: Steps to arrive at a tissue analysis

This experiment describes the steps to arrive at a tissue analysis using high wavenumber -Raman spectroscopy The steps may be implemented in various ways (the description of the steps below, are therefore given by way of example and are not meant to be limiting in any way):

1) Tissue is illuminated through an optical fiber and light which is scattered by the tissue is collected by the same optical fiber.
2) The Raman spectrum of the collected light is recorded in the form of signal intensities vs. detector channel number.
3) The measured spectrum is pre-processed prior to final analysis, this pre-processing step may comprise wavenumber calibration of detector channels, correction for varying wavenumber–dependent signal detection efficiency, correction of measured spectra for background signal contributions, generated anywhere in the Raman measurement system, but not due to the tissue under investigation.
4) Analysis of the pre-processed spectra. As an example a classical least squares analysis may be used in which the measured spectrum is fitted with spectra of compounds of which it is known that they may be present in the tissue in amounts sufficient to have a detectable contribution to the overall tissue spectrum and e.g. a polynomial with coefficients that can also be fitted to optimally take account of slowly varying backgrounds to the Raman spectrum that may be due to e.g. fluorescence excited within the sample. When the compound-spectra are intensity-scaled prior to fitting the tissue spectrum, in such a way that the fit-coefficients for compound-spectra resulting from a fit of a spectrum of a sample containing equal amounts of these compounds, would be equal, then, apart from the fact that in practice different efficiencies may apply for collection of signal from different tissue volumes and that the tissue may be heterogeneous in molecular composition, values of the fit-coefficients are directly related to the weight-percentages of the respective compounds present in the tissue on condition that the tissue is sufficiently homogeneous. If this is not the case, the composition as determined will still be in qualitative agreement but not necessarily in quantitative agreement with the real composition. For instance, because the arterial wall and atherosclerotic plaque are not homogeneous in molecular composition, and because depending on probe geometry, Raman signal is collected with different efficiencies from different

tissue volumes, and because of signal attenuation within the tissue, certain tissue volumes, with potentially different molecular composition, will contribute signal more effectively than others. The weight percentages of compounds present in the tissue may represent the actual information sought, or they may be used to type the tissue and determine its clinical diagnostic class. Alternative approaches for determining weight percentages of specific compounds or groups of compounds include the well known partial least squares analysis. Also other multivariate statistical signal analysis approaches such as principal components analysis, linear discriminant analysis, logistic regression analysis, or e.g. analysis based on an artificial neural network may be applied for determining the clinical diagnostic class of a tissue.

5) Outputting the desired data in a visible or audible form as well as storing the data with proper references for future assessment and/or cross-referencing with other data, such as e.g. coordinates of the location of measurement, or images of the location where the Raman spectrum was measured, e.g. an angiogram or intravascular ultrasound images.

Example 6: Lipid measurements

Figure 7 shows a spectrum (A) of a mixture of lipids measured with a Raman setup according to Figure 6. Specifically, the laser 100 was a ti-Sapphire-laser (model 3900S, Spectra Physics, USA) emitting laser light at 720nm. Filter 140 was a custom made dielectric filter (produced by Omitec, UK) which transmitted laser light of 720nm and which reflected light returned from the sample with a wavelength above 850nm. The direction of the incoming laser light and the normal to the filter surface included an angle of 15 degrees. Lens 110 was a microscope objective for use in the near-infrared (x20 PL-FL Nachet, numerical aperture 0.35). The optical fiber 120 was a WF200/220A optical fiber from Ceramoptec. The filter 150 was a color glass filter RG 780 (Schott). Light transmitted by filter 150 was imaged onto a an optical fiber with a core of 1000 microns which was connected to a round bundel of 64 optical fibers with a core diameter of 100 microns. At the distal end of this bundle the fibers were arranged in a linear array and light was guided into spectrometer 160 in this way. Spectrometer 160 was a Renishaw system RA 100 imaging spectrometer equipped with a deep-depletion CCD-camera for multichannel signal detection. Also shown are the spectrum

of the fiber optic probe itself (B, obtained without a sample present at the distal end of the optical fiber) and a difference spectrum A-B, illustrating that with a single properly selected unfiltered optical fiber, high quality spectra can be obtained of samples of similar molecular composition as may be encountered in atherosclerotic lesions.

Figure 8 shows Raman spectra (t) of a normal artery wall (A) and an atherosclerotic artery wall (B), of the results (f) of a least squares fitting of these spectra with the set of spectra of purified compounds shown in figure 2 and of residuals (r) which represent the signal contained in the tissue spectra that is not accounted for by the set of fit-spectra. As can be seen by the low intensity of the fit-residuals, the fit of the tissue spectra is highly accurate enabling detailed information regarding molecular composition of the tissues to be obtained. This result is shown by way of example. For instance, the set of compound spectra which is used to fit the tissue spectra, may be composed of other spectra or a different number of spectra.

Table 2 shows a table with weight percentages of compounds or compound groups of the arterial samples of which the spectra are shown in figures 8A and 8B, as determined from the results of the least squares fit analysis. The spectrum of the normal artery is dominated by signal contributions of triglycerides, representing the adventitial fat signal contributions, no or very minor signal contributions from cholesterol and cholesterol esters are found, in contrast with the signal obtained from the atherosclerotic artery which contains significant signal contributions from cholesterol and cholesterol esters.

TABLE 2: weight percentages of compounds or compound groups of the arterial samples of which the spectra are shown in figures 8A and 8B

	Normal artery	**Atherosclerotic artery**
Cholesterol linoleate	0.0078	0.3776
Cholesteryl oleate	0	0.0532
Cholesteryl linolenate	0.02	0
Cholesteryl palmitate	0.0187	0.1155
Trilinoleine	0.0477	0.0235

	Normal artery	Atherosclerotic artery
Triolene	0.7937	0.1436
Tripalmitine	0.0032	0
Cholesterol	0	0.1530
Collagene	0.0633	0.0756
Elastine	0.0456	0.0574

This experiment shows that spectrometer of the invention enables Raman spectroscopy as in vivo technique for studying artherosclerotic plaque, but now with the above mentioned benefits of this spectrometer.

Example 7: Instrument with fibers measuring fluorescence and or NIR absorption

Figure 9 shows schematically an embodiment in which Raman spectroscopy is combined with fluroscence and NIR-absorption spectroscopy. This embodiment shows one single fiber on the left side of the figure and excitation light that is coupled via reflectors into the fiber. The same or another reflector is used to decouple from the obtained signal out of the fiber the fluorescence light for detection. Further to the right, another reflector couples laser light into the fiber for producing a Raman signal from a sample. The same or another reflector is used to decouple the Raman signal out of the fiber to a detector. On the right hand side of the figure, NIR light of a NIR source is coupled into the fiber, and the NIR signal that is guided back by the same fiber is measured by a suitable detecor. Meaurements can be done sequentially or simultaneously. The fiber shown can also be a bundle of fibers. The person skilled in the art will adapt the optics, sources, detection units etc. to his purpose, the tissue to be measured or the information that is desired.

Whilst specific embodiments of the invention have been described above, it will be appreciated that the invention may be practiced otherwise than as described. For example, the instrument can also be used to measure biological molecues, like lipids, etc. in other species than tissue, e.g. for use in the analysis of milk, oil, etc. The description and the examples are not intended to limit the invention.

Claims

1. Use of an instrument for measuring a Raman signal of tissue, the instrument comprising a laser, a signal detection unit for measuring the Raman signal, and a fiber optic probe, wherein the fiber optic probe comprises one or more optical fibers for directing laser light onto the tissue and for collecting light that is scattered by the tissue and guiding the collected light away from the tissue towards the signal detection unit, wherein the fiber or fibers for collecting light have substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region, and wherein the detection unit records the Raman signal scattered by the tissue in said spectral region.

2. Use of an instrument according to claim 1, wherein the fiber optic probe comprises an optical fiber that both directs laser light onto the tissue and collects light that is scattered by the tissue and guides the collected light away from the tissue towards the signal detection unit, and wherein the fiber has substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region.

3. Use of an instrument according to claims 1 or 2, wherein the detection unit records the Raman signal in one or more parts of the 2700-3100 cm-1 spectral region.

4. Use of an instrument according to one of claims 1-3, wherein the fiber optic probe comprises an optical element at the distal end of the optical fiber.

5. Use of an instrument according to one of claims 1-4, further comprising a signal analysis unit which analyses the recorded Raman signal, the analysis comprising an algorithm which outputs data regarding the molecular composition of the tissue and/or the clinical diagnostic class to which the tissue belongs.

6. Use of an instrument according to claim 5 for analysis and/or diagnosis of atherosclerotic plaque.

7. Use of an instrument according to claim 5 for use in the analysis and/or diagnosis of cancerous tissue or pre-cancerous tissue.

8. Use of an instrument according to one of claims 1-7, wherein the fiber optic probe is integrated or combined with a catheter.

9. Use of an instrument according to one of claims 1-8, wherein Raman measurements can be combined with fluorescence and/or near-infrared absorption measurements and wherein the detection unit also comprises a detection unit for measuring the intensity and/or spectrum of tissue fluorescence and/or a detection unit for measuring near-infrared absorption.

10. Use of an instrument according to claim 9, wherein fluorescence and/or near-infrared absorption measurements make use of a fiber also used in obtaining Raman signal.

11. Use of an instrument according to one of claims 1-10, wherein the fiber optic probe comprises a bundle of fibers for measuring and/or scanning a tissue area.

12. Use of an instrument according to one of claims 1-10, wherein the fiber optic probe comprises one optical fiber, the fiber having substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region.

13. Use of an instrument according to one of claims 1-12, wherein the fiber optic probe is brought in, or in contact with, or in proximity to the tissue under investigation.

14. Use of an instrument according to one of claims 1-13, wherein the tissue is excised, biopted or taken from a human or animal body before measuring.

15. An instrument for measuring a Raman signal of tissue, the instrument comprising a laser, a signal detection unit for measuring the Raman signal, and a fiber optic probe, wherein the fiber optic probe comprises one or more optical fibers for directing laser light onto the tissue and for collecting light that is scattered by the

tissue and guiding this collected light away from the tissue towards the signal detection unit, wherein the fiber or fibers for collecting light have substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region, and wherein the detection unit is able to record the Raman signal scattered by the tissue.

16. An instrument for measuring a Raman signal of tissue according to claim 15, wherein the fiber optic probe comprises an optical fiber that both directs laser light onto the tissue and collects light that is scattered by the tissue and guides this collected light away from the tissue towards the signal detection unit, and wherein the fiber has substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region.

17. An instrument according to claims 15 or 16, wherein the detection unit also comprises a detector for measuring fluorescence and/or a detector for near-infrared absorption.

18. An instrument according to claim 17, wherein fluorescence and/or near-infrared absorption measurements make use of a fiber also used in obtaining Raman signal and wherein the detection unit also comprises a detector for measuring fluorescence and/or a detector for near-infrared absorption.

19. An instrument according to one of claims 15-18, wherein the fiber optic probe comprises a bundle of fibers for measuring and/or scanning a tissue area.

20. An instrument according to one of claims 15-18, wherein the fiber optic probe comprises one optical fiber, the fiber having substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region.

21. Method for measuring a Raman signal of a tissue sample, wherein an instrument according to one of claims 15-20 is used and wherein the tissue sample is excised, biopted or taken from a human or animal body before measuring.

Abstract

The invention is related to the use of an instrument for measuring a Raman signal of tissue, the instrument comprising a laser, a signal detection unit for measuring the Raman signal, and a fiber optic probe, wherein the fiber optic probe comprises one or more optical fibers for directing laser light onto the tissue and for collecting light that is scattered by the tissue and guiding the collected light away from the tissue towards the signal detection unit, wherein the fiber or fibers for collecting light have substantially no Raman signal in one or more parts of the 2500-3700 cm^{-1} spectral region, and wherein the detection unit records the Raman signal scattered by the tissue in said spectral region. The invention enables *in vitro* and *in vivo* analysis and diagnosis of atherosclerotic plaque and detection of tumor tissue with great advantages over current state-of-the-art technology.

Fig 1a

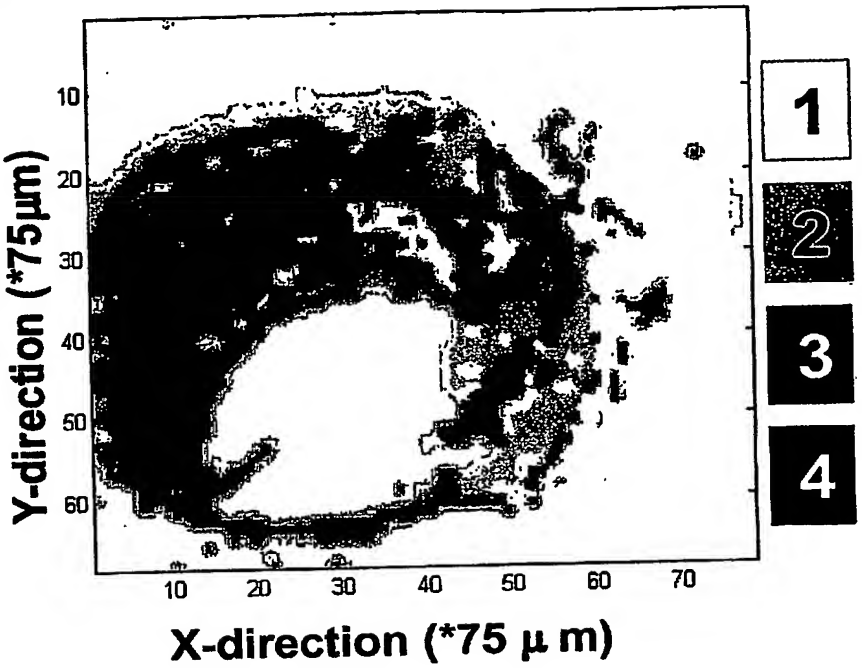

Y-direction (*75µm)
10
20
30
40
50
60
10
20
30
40
50
60
70
X-direction (*75 µ m)
1
2
3
4

Fig 1b

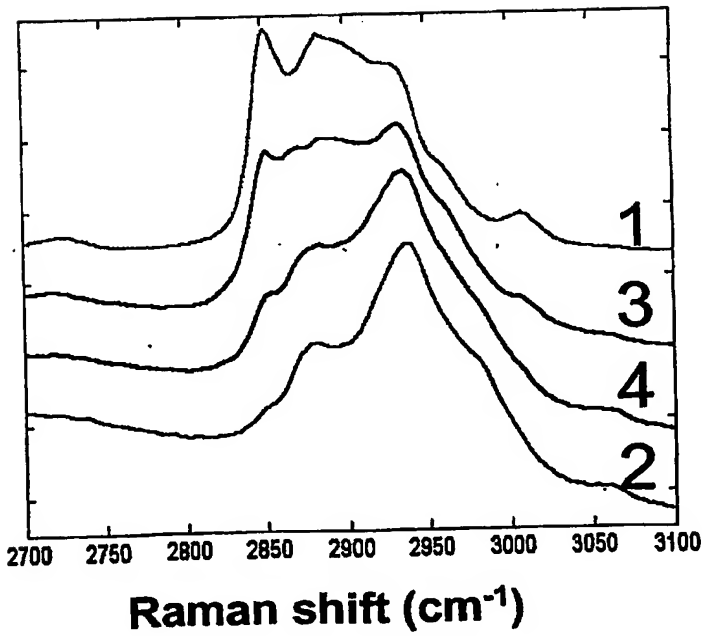

1
3
4
2
2700
2750
2800
2850
2900
2950
3000
3050
3100
Raman shift (cm-1)

Fig 2

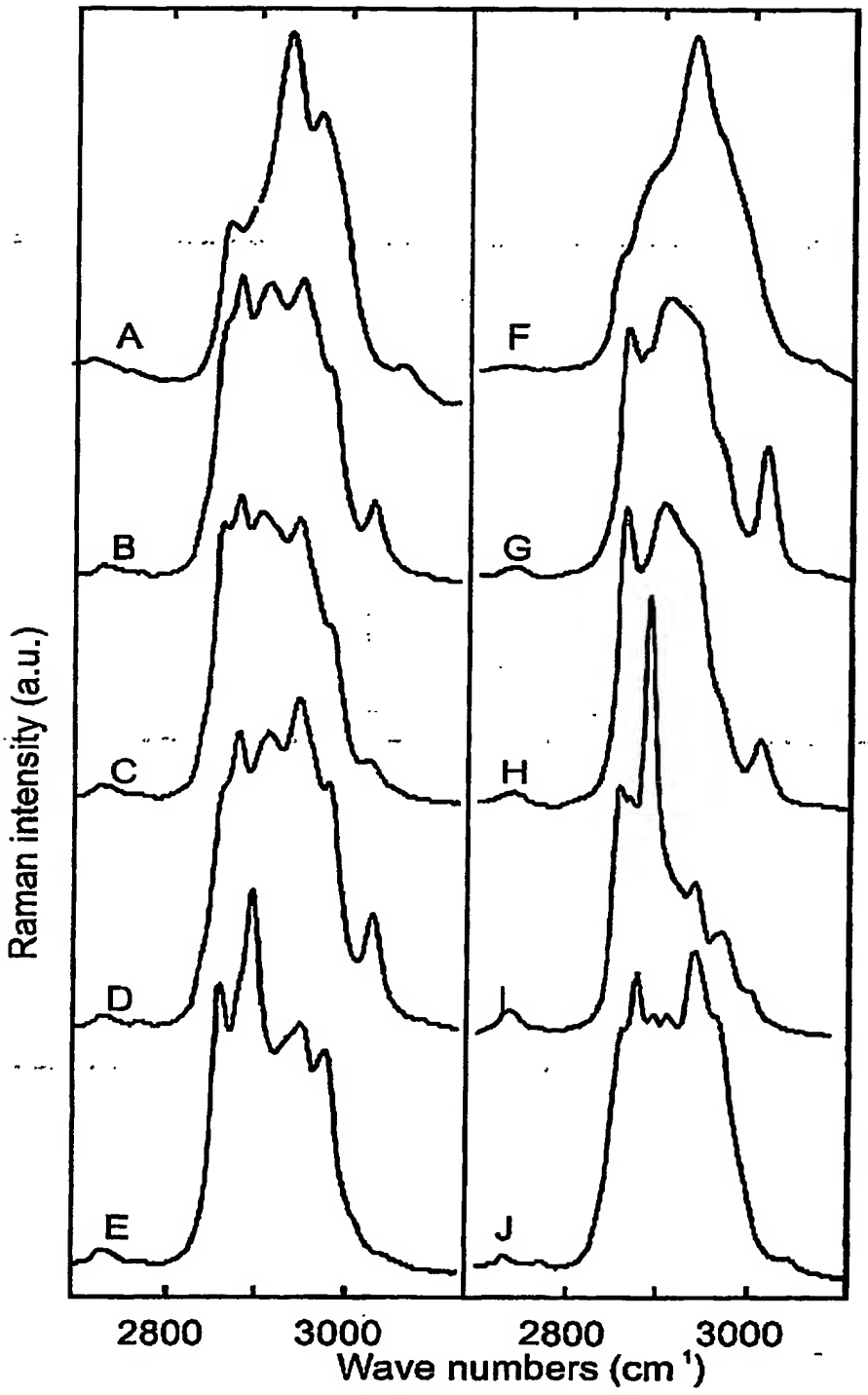
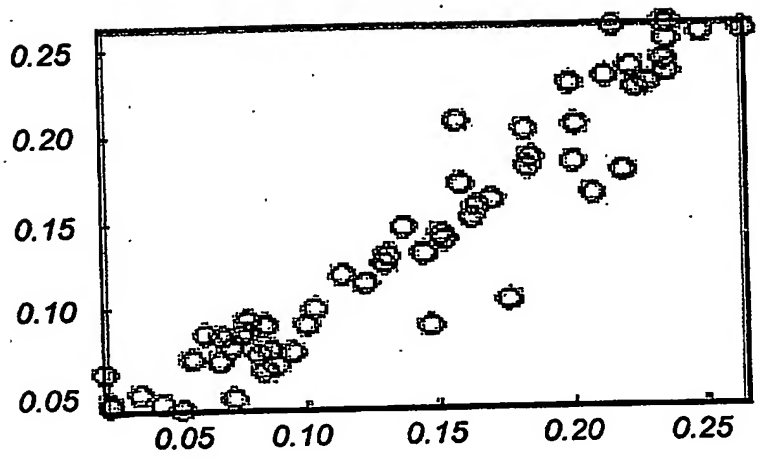
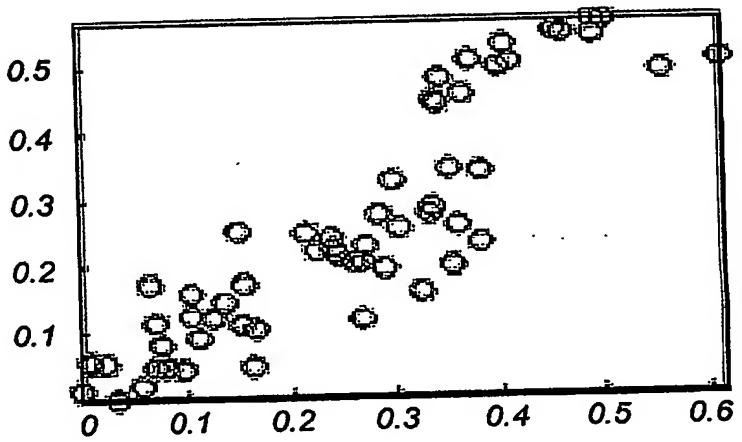

A
B
C
D
E
F
G
H
J
Raman intensity (a.u.)
2800
3000
2800
3000
Wave numbers (cm⁻¹)

Fig 3

Cholesterol
R=0.95



Cholesteryl
esters
R=0.93



triglycerides
R=0.96

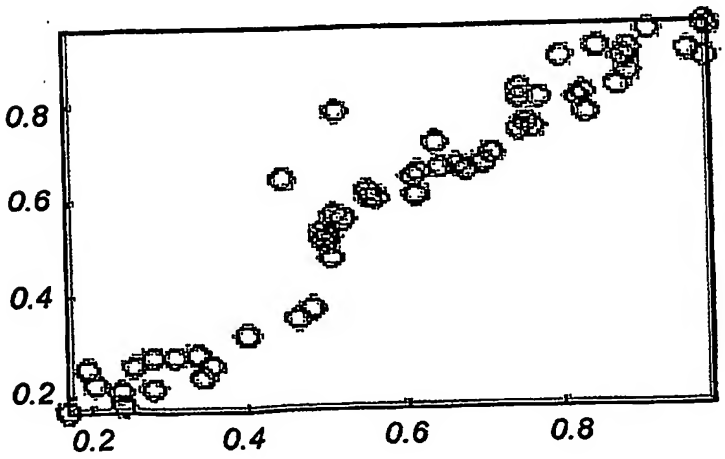


Fig 4a

Raman map

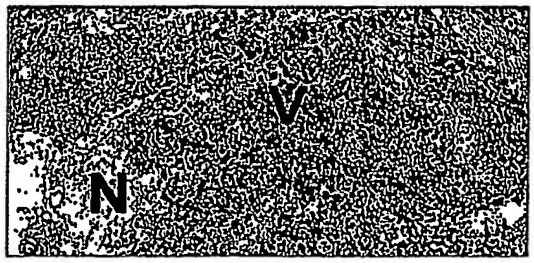
Fig 4b

adjacent section,
H&E stained

Fig 5a

K-means pseudo-
color **Raman map**

Fig 5b



adjacent section,
H&E stained

Fig 6

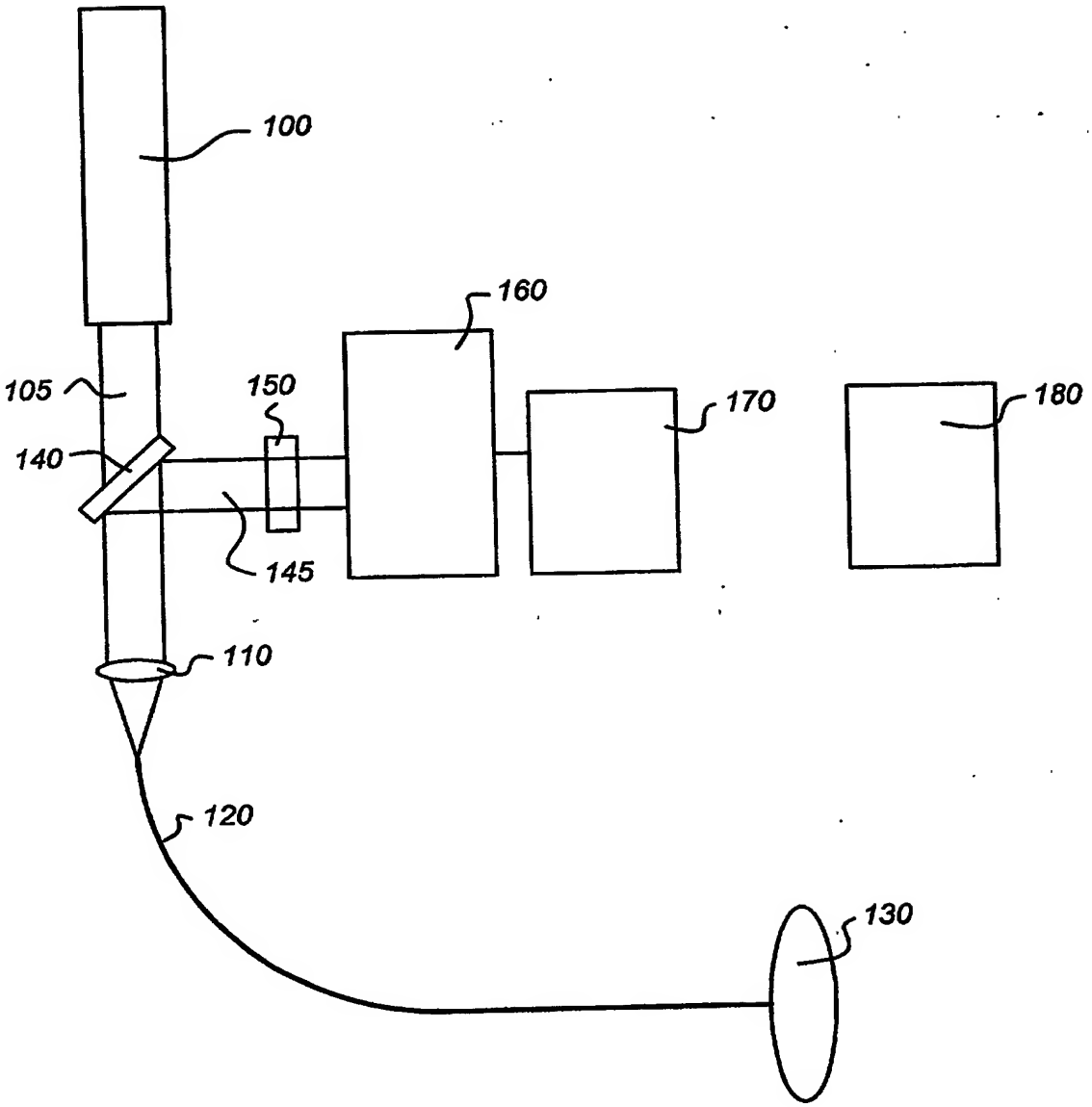


Fig 7

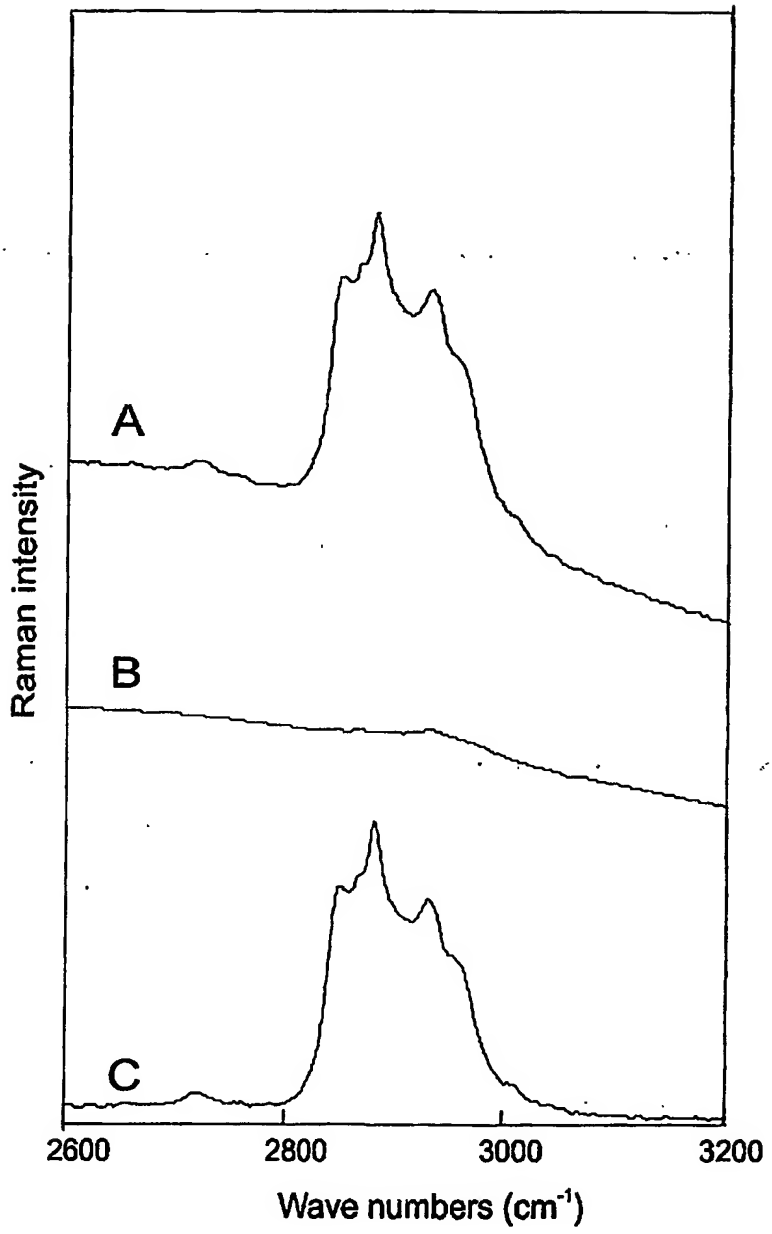


Fig 8a

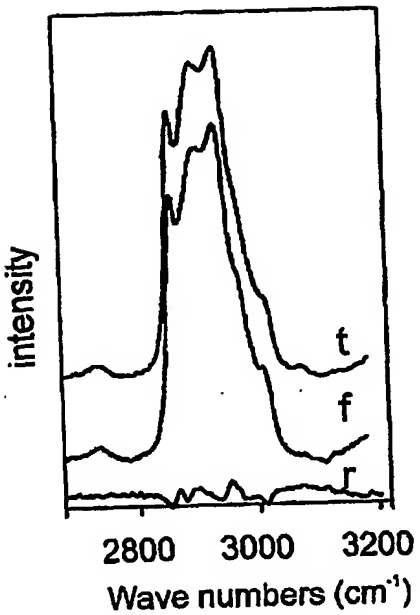
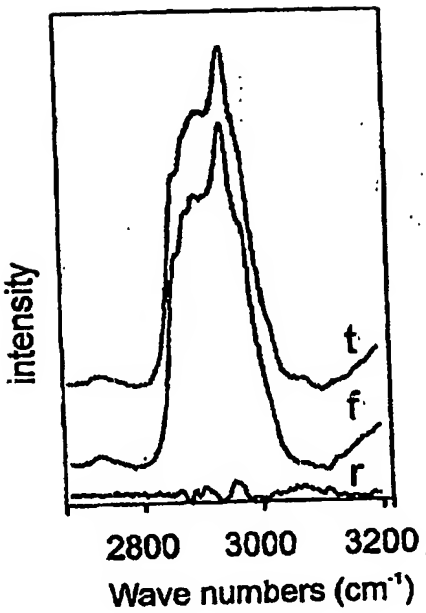

intensity
t
f
r
2800
3000
3200
Wave numbers (cm-1)

Fig 8b


intensity
t
f
r
2800
3000
3200
Wave numbers (cm-1)

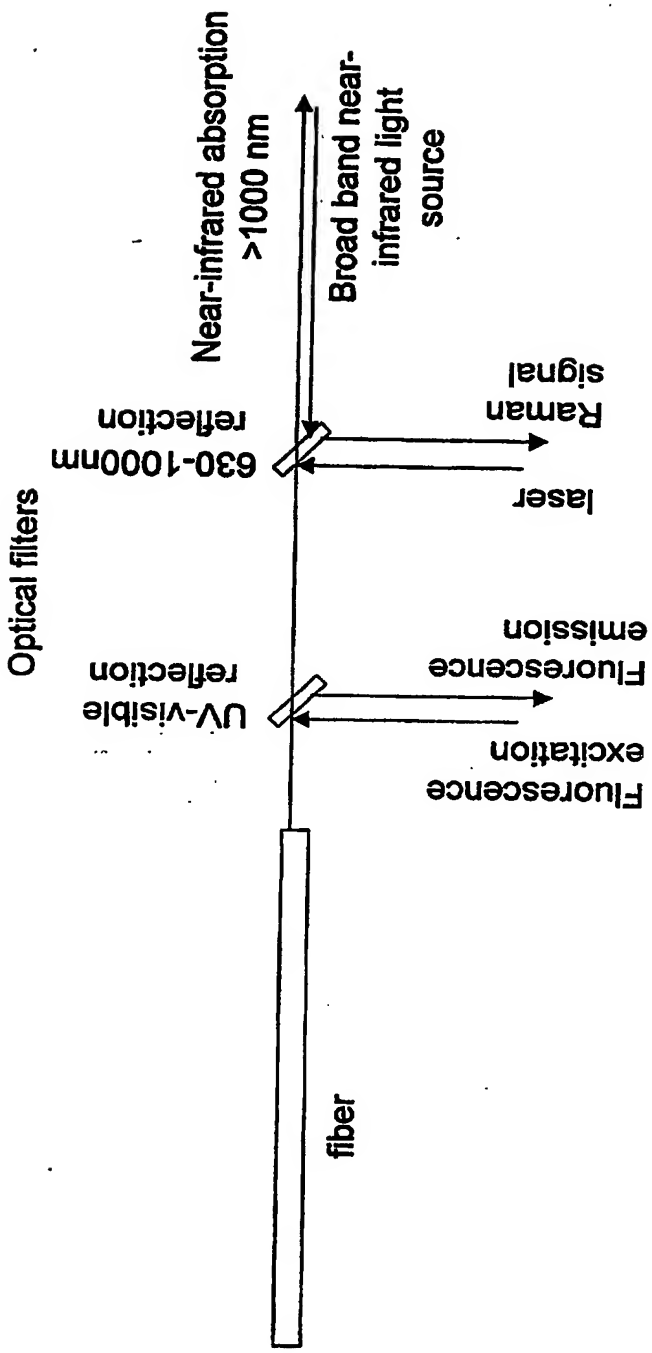

Near-infrared absorption
>1000 nm
Broad band near-infrared light source
Raman signal
630-1000nm reflection
laser
Optical filters
UV-visible reflection
Fluorescence emission
Fluorescence excitation
fiber

Fig 9